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Research article

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Kinetic energy scavenging in a prosthetic foot using a fluidic system

Abstract: The use of active prostheses for the lower extremity is limited by the amount of electric energy stored in batteries. A promising way to extend their usage time is to convert motions generated by the human body during walking to electrical energy. A first functioning prototype was designed to transfer kinetic energy from heel contact and forefoot contact to a generator by using a fluidic system. Experimental results show that walking with the system generates an average electrical power of 0.8 W. The design of the energy scavenging system (ESS) is presented and results are discussed.

Keywords: biomechanics; body energy harvesting; fluidic system; foot prosthesis.

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Introduction

In recent years, a number of active prostheses were developed for the lower extremity, e.g., the Biom model by iWalk (www.iwalkpro.com) the C-Leg® from Otto Bock healthcare (www.ottobock.com) or the Proprio-Foot and Rheo-Knee by Össur (www.ossur.com). Compared to passive prostheses, their functionalities were extended: They can adapt to various terrains and slopes and have an increased stability, e.g., when climbing stairs. As a result, healthy extremities can be relieved. Due to their various functions, active prostheses require electric energy, which is typically supplied by a battery. The battery has to be recharged regularly and it has a considerable share in the total weight of the system. To extend the service life of an active leg prosthesis, the body's own energy might be converted into electric energy during walking.

A walking cycle starts with the first ground contact at the heel when the foot is put down and ends with the following ground contact of the same foot and is divided in stance and swing phases. The walking cycle comprises mainly four different energy-rich motions:

1. Heel contact and 2. forefoot contact: During the stance phase of one foot, two load peaks of vertically directed ground reaction forces occur, corresponding to 120–150% of the body weight [1]. The first peak occurs during contact of the rear third of the foot with the ground (heel contact), whereas the second peak is related to pushing off with the forefoot at the end of stance (forefoot contact). Prior to the heel contact, a smaller load peak is observed, which results from the sudden compression of the connective tissue plate of the heel.
3. Center of mass motion. During walking, the center of mass of a person moves up to 50 mm in vertical direction [17]. If an external mass (e.g., a backpack) is forced to follow the motion of the center of mass, a considerable amount of mechanical energy has to be applied. Only if the mass is decoupled from the body, such that a relative movement between mass and body occurs, mechanical energy can partly be converted into electrical energy. According to [12], about 1 W can be produced per kg mass. However, application of an additional load due because of an energy harvesting device increases user effort and therewith the metabolic cost [7]. The muscles would have to perform extra work to move the additional weight of the device.
4. Joint motion. During walking, the muscles in the arm and leg joints produce torques that vary in amount and direction as a function of the progress of step motion. If the muscles slow down the extremities, a negative torque is acting, and vice versa. In [17], joint position, rotation speed, torque, and power of the ankle, knee, and hip were measured and classified into positive and negative muscle work. Based on these data, [12]

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determined the negative power that can be produced at the joints for a person of 80 kg body weight. Highest torques and negative power act in the leg joints, in particular in the ankle with 140 Nm and 19 W. The phases, in which the leg muscles produce negative power, are sufficiently long to extract a considerable amount of energy. As negative and positive work is accomplished alternately, one challenge is using solely the negative power to scavenge energy. Otherwise, extra metabolic work has to be performed to drive the generator.

State of the art and requirements on an energy scavenging system (ESS)

To convert the body's own energy into usable electric energy during walking, various energy conversion systems can be applied. In the past, systems with linear generators integrated into the sole were presented [3, 9, 13, 20]. The generator power depends on the stimulation frequency and they generate up to 674 mW. However, further minimization of the most powerful design [20] is required to allow for integration into a prosthetic foot. Electromagnetic generators, based on a rotatory principle were applied [4, 6, 7]. The challenge of this principle is to convert biomechanical power with low speed and high torque into mechanical power with high speed and low torque and to transmit it to the generator with low losses and high energy efficiency. For this purpose, a generator mechanism from a hand-powered torch was used with a flywheel driven by a lever that is compressed by 3 cm when putting down the foot. Average electric powers of 250 mW were generated [6]. In addition, two patents [18, 19] cover systems consisting of two deformable liquid reservoirs in the heel and ball area of the shoe and a turbine coupled to a generator. A pressure difference between the inlet and outlet of the turbine caused by the compression of one of the reservoirs is compensated by medium flow, which makes the turbine rotate and supply electric energy. Apart from electromagnetic systems, shoes with integrated piezoelectric or capacitive generators were presented [6, 2, 10, 14]. The PZT stack generators are relatively robust, but also rather expensive and the systems generate a few mW only. Moreover, prototypes with electrostatic generators [11, 15] were built, but not further developed. None of the systems presented explicitly serves to supply an active leg or foot prosthesis with electricity.

An ESS for leg or foot prostheses has to meet the following criteria.

Power

The system has to supply sufficient electric energy for supporting an active prosthesis (such that batteries have to be recharged less often). A value of 1 Ws/step is targeted. This energy is to be stored and supplied by the system with minimum losses [16].

Weight and additional metabolic expenditure

The system will only be advantageous for the user and accepted by him if he does not have to actively provide the electric energy, also known as parasitic energy harvesting. That means that the electricity is to be based on the energy that has to be applied by the user anyway, but has not been used so far. For this reason, the system should be as light as possible. The additional weight of about 1 kg of active prostheses compared to passive prostheses can be used as an orientation value.

Space available

To prevent the system from being annoying in everyday life, e.g., when wearing shoes, a space of about 50×50×150 mm only is available in the rear area of the prosthesis.

Ergonomics

Walking with a prosthesis equipped with an ESS should hardly differ from walking without an ESS. Noise should be minimized.

Independence of energy supply

For using the system, no additional electric energy source should be required. This means that supply of components, such as valves or actuators, with an external power supply should be avoided.

Service life

Service life of the system should correspond to that of a prosthesis of about 2 years or 2 million cycles. Failure of the system should not affect the motion sequence of the user and in the case of a failure, individual components should be easily exchangeable.

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Risk potential

In accordance with the requirements outlined in the Medical Devices Act, any potential danger for the user in any state of operation of the system has to be prevented. Consequently, high currents and voltages have to be avoided and shielded by a housing and movable components have to be encapsulated.

System components

The ESS was developed on the basis of a passive foot prosthesis for active patients (mobility degree 3–4), Vari-Flex model by Össur, Reykjavik, Iceland, size 28 (corresponding to a shoe size of 45). It is envisaged that the ESS will be integrated in a powered foot-ankle prosthesis at a later date. In contrast to prosthetic knee joints, foot prostheses are characterized by deformation of a glass-fibre reinforced plastic (GRP) spring without rotation in a hinge joint during walking. To convert kinetic energy, both heel contact and forefoot contact must be used. The relatively small motion is transmitted via flexible fluid reservoirs and a differential cylinder to a ball screw. This ball screw drives a gear that drives a rotatory electromechanical generator of high rotation speed. To save costs, existing standard components are used. The components of the ESS are illustrated in Figure 1.

1. Fluid system: the fluid system consists of a fluid cushion in the heel and ball area each, which are compressed alternately during walking. The flexible fluid reservoirs [16] have a diameter of 40 mm and a volume of about 12 ml at a deformation of 15 mm. The fluid used is distilled water that is passed to a short-stroke differential cylinder made by Festo (type ADVC-25-25-A-P).
2. Gear: by a connection element, the piston movement of the differential cylinder is transmitted to a ball screw that converts linear movement into rotation movement. A standard ball screw (type KGT ED 513X/V501X by Faulhaber, Schönaich, Germany) was chosen. From an axial load of about 35 N, it reaches an efficiency of 87%. The rotional speed was doubled by using a second gear with a ratio of 1:2.
3. Electric generator: selection of the electric generator represents a compromise of weight, internal resistance, and output voltage. The latter depends on the rotation constant and the rotation speed. A permanent rare-earth magnet DC electric motor Type 221012 of the Re-max series by Maxon Motors, Sachseln, Switzerland was chosen. At a target rotational speed of 4000 revolutions per minute, this motor has an output voltage (U) of 3.67 V when used as a generator with a load resistance (R_L) of 30 Ω. Its electrical power output (P) of 449 mW can be calculated by the term $P=U^2/R_L$.
4. Energy conversion system: to store the strongly fluctuating sinusoidal electric alternating voltage of the generator, rectification, smoothing, and conversion to a voltage level compatible to a storage medium are required. The company Linear Technologie offers special energy harvesting DC/DC buck convertors (type LTC3588-1 by Linear Technology, Milpitas, CA, USA). The electronic circuit is given in Figure 2A and B. Energy flow to and from the energy storage system should then be controlled by an energy management system [5].

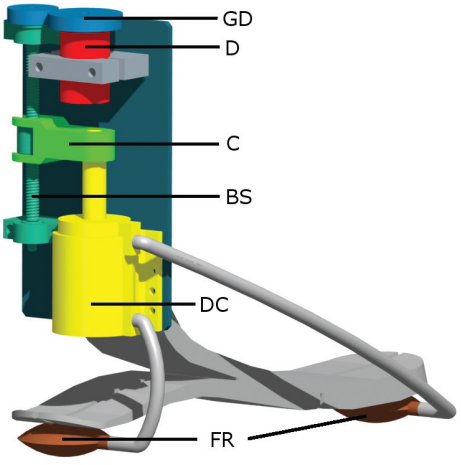

Figure 1 Components of the ESS: EP, energy processing (not illustrated); G, generator; BS, ball screw; DC, differential cylinder; FR, fluid reservoir; GD, gear drive; C, connector.

Evaluation of the ESS

The ESS developed was evaluated under laboratory conditions. A healthy test person of 80 kg weight walked with the ESS on a concrete floor at an average cadence of 0.77 steps/s. A video camera recorded the stance phase with a rate of 30 frames/s. For all measurements, an isolated multi-functioning I/O data acquisition device (type NI USB-6215 by National Instruments Inc., Austin, TX, USA) was used. The scanning rate was 1.500 samples/s. Data were recorded using LabView™ software and processed using Excel software from Microsoft. For evaluation of the electric power output of the ESS, the fluid cushions were bonded to the lower part of the carbon-composite foot prosthesis at the heel and the forefoot area (see Figure 3).

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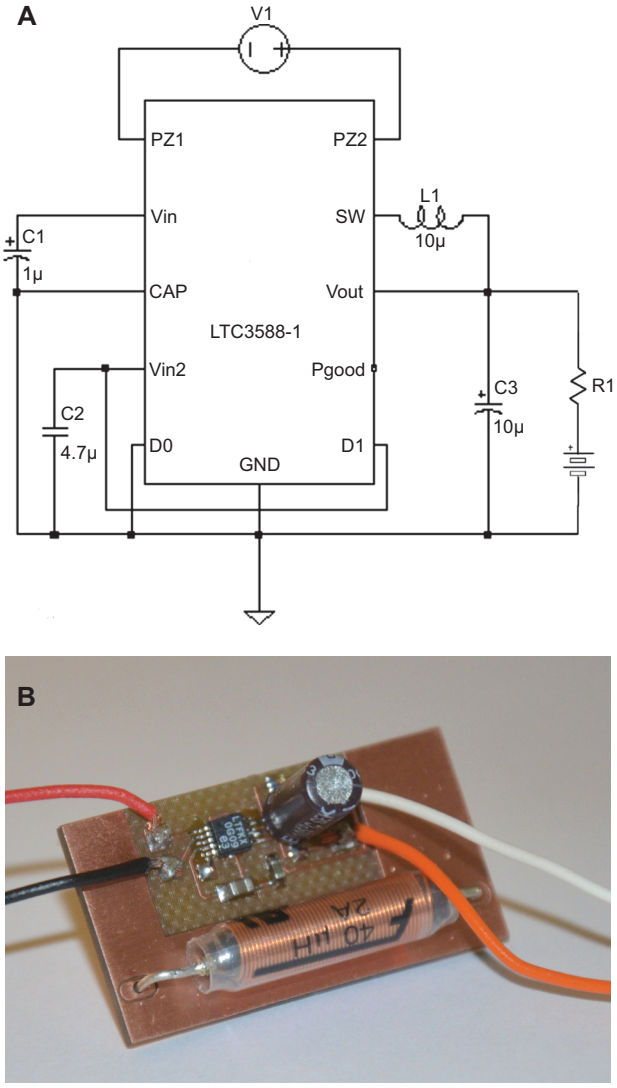

Figure 2 (A) Electronic circuit and (B) circuit diagram of the energy conversion system.

The upper part of the carbon-composite foot prosthesis was replaced by a cross country running shoe that was reinforced by an aluminum plate of 8 mm thickness and which was connected to the lower part of the prosthesis.

Additionally pressure sensors (GD14HN, Greisinger Electronics, Regenstauf, Germany) were connected to the pressure lines of each fluid cushion and data were recorded synchronously with the output voltage of the generator. The median values for ten stance phases were calculated.

Results and discussion

Video analysis revealed that the average single-leg stance phase of ten measurements was 0.79 s. This corresponds to a stride duration of 1.29 s and to cadence of 0.77 steps/s when the duration of stance is 61% of the walking cycle, as given in [8]. On average the fluid cushion at the heel region is drained within 0.27 s and at the forelimb within 0.37 s, respectively. Further acceleration of fluid cushion draining can mainly be achieved by using pressure lines with a larger diameter. In addition, the shape of the fluid cushions has potential for optimization (e.g., the central deformation of 15 mm can be reduced by using a fluid cushion with toroidal shape).

As depicted in Figure 4, the output voltage curve has the same characteristic as the two pressure curves. The delay in the output voltage in comparison to the pressure curves results from inertia in the pressure build-up, and also from the minimum speed of generator rotation that is necessary to produce an output of electric power. The pressure curve has two maxima during heel contact. Using video analysis it can be identified that the first maximum

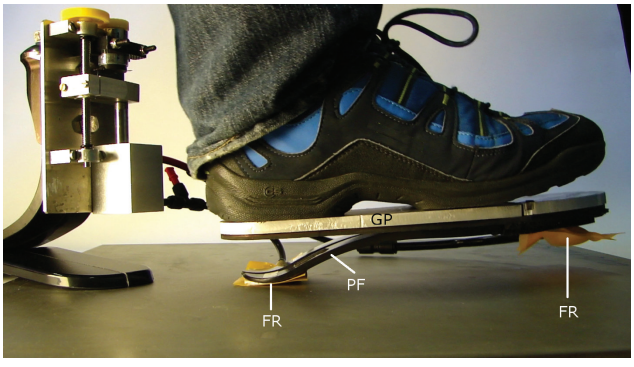

Figure 3 Experimental setup of the evaluation. The picture-excerpt from the video analysis corresponds to the end of fluid cushion (FC) deflation at the heel. An aluminum ground plate (GP) is connected to a sports shoe and to the lower part of a prosthetic foot (PF).

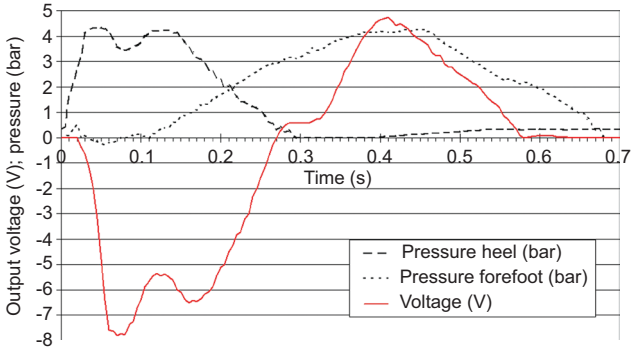

Figure 4 Pressure behavior in the two fluid cushions and output voltage of the generator during the stance phase of a single step. The displayed value curves correspond to data of a single measurement. The time=0 s is shortly before heel-strike and the figure shows the time from heel-strike to toe-off.

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occurs during the start of heel contact. Furthermore, it can be assumed that the elastic fluid cushion is dilated and the stored mechanical energy is released during the second maximum. During forefoot contact the pressure in the elastic fluid cushion increases linearly until it remains on a plateau level for approx. 0.15 s. The highest pressure in the system is 4.3 bar.

An alternating voltage of up to 9 V with a frequency that depends on the cadence can be measured at the generator output. In the example given in Figure 4, the maximum generator output voltage is 7.7 V and the fluid cushions are drained within 0.23 and 0.26 s, respectively. The real electric power (P) averaged over a period of time (T) can be obtained by taking the integral of the instantaneous calculation of the product of the voltage $u(t)$ and current $i(t)$ over the desired period as given in Eq. (1).

$$P=\frac{1}{T_{period}}\int_{t_1}^{t_2} u(t)\times i(t)\,dt \tag{1}$$

Measurements revealed that the system delivers an average electric power of up to 797 mW at a cadence of 0.77 steps/s. The average electric power output during heel contact was rated at 1.12 W, whereas during forefoot contact it was 0.67 W, respectively. One explanation for this is that the pressure line of the front cushion is longer, resulting in a higher friction. The power of 0.8 W corresponds to a volumetric power density of 6.28 mW/cm^3 considering a volume of 127 cm^3 of the ESS. A comparison of the volumetric energy density of the system developed with existing shoe-based systems clearly revealed the advantages and good performance of the system (see Table 1). By integrating a ball screw in the differential cylinder, total weight of the system can be further reduced and an energy density of >20 mW/cm^3 can be reached.

After rectification and smoothening the alternating voltage is converted into a constant voltage of 3.3 V, e.g., to recharge an accumulator, by the use of a DC-DC buck-boost converter (LTC3588-1 from Linear Technologies, Milpitas, CA, USA). After a short period of delay (0.1–0.2 s) where the capacitors of the circuit have to be charged, a constant voltage of 3.3 V can be measured. The current shows a pulsed behavior and suggests losses during energy processing, which were not studied in further detail yet.

Table 1 Comparison of different power conversion principles of shoe-based systems.

Principle	Power (mW)	Volume (cm^3)	Power density $\left(\frac{mW}{cm^3}\right)$
Own solution	797	127	6.28
Piezoelectric (PVDF) [14]	1.3	5	0.26
Piezoelectric (PZT) [14]	8.4	1.7	4.94
EAP [11]	1000	50	20
Electromagnetic (linear) [20]	674	200	3.4
Electromagnetic (rotary) [14]	60	56	1.07

Summary and outlook

Principle performance and functionality of a combined fluidic/electromagnetic system for energy scavenging during walking with a prosthetic foot were demonstrated. DC generators without brushes were not compared with the tested generators with brushes, as their circuit complexity is much higher. Circuit measurements confirmed the simulated result of a constant output voltage. In this respect, it was succeeded in converting the energy produced into a form compatible with conventional energy storage systems. As postulated in the requirements, no additional electric energy source is needed to operate the ESS. The target of scavenging 1 W of electric power per step was reached by 80% and is already sufficient to considerably reduce the frequency of recharging an active prosthesis. The prototype of the ESS can already be housed in the rear of a prosthetic foot. However, there is potential for a further reduction of the systems volume and overall weight by reducing the number of components (e.g., by integration of the ball screw in the differential cylinder). Additionally, an evaluation of the metabolic energy consumption with and without ESS is required. The own weight of the ESS causes an additional metabolic consumption that should be as small as possible [7, 12]. Service life of the system also must be evaluated and all components have to be optimized to guarantee operation of 2 years.

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